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Effects of vision and lumbar posture on trunk neuromuscular control

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ABSTRACT

The goal of this study was to determine the effects of vision and lumbar posture on trunk neuromuscular control. Torso perturbations were applied with a pushing device while the subjects were restrained at the pelvis in a kneeling-seated position. Torso kinematics and the muscle activity of the lumbar part of the M. Longissimus were recorded for 14 healthy subjects. Four conditions were included: a flexion, extension and neutral lumbar posture with eyes closed and the neutral posture with eyes open. Frequency response functions of the admittance and reflexes showed that there was no significant difference between the eyes open and eyes closed conditions, thereby confirming that vision does not play a role in the stabilization of the trunk during small-amplitude trunk perturbations. In contrast, manipulating posture did lead to significant differences. In particular, the flexed condition led to a lower admittance and lower reflex contribution compared to the neutral condition. Furthermore, the muscle pre-activation (prior to the onset of the perturbation) was significantly lower in the flexed posture compared to neutral. This confirms that flexing the lumbar spine increases the passive tissue stiffness and decreases the contribution of reflex activity to trunk control.

1. Introduction

The human trunk is inherently unstable without motor control, as any deviation from the vertical will be aggravated by gravity. Research into how unstable postures are controlled has mainly focused on the control of upright stance, often considering the body as a single inverted pendulum rotating around the ankles (van der Kooij et al., 2001; Welch and Ting, 2008). However, understanding the stabilization of the trunk specifically might be important as impaired trunk control has been observed in patients with such diverse disorders as low back pain (Descarreaux et al., 2005; Willigenburg et al., 2013), Parkinson's disease (van der Burg et al., 2006), stroke (Ryerson et al., 2008) and spinal cord injury (Seelen et al., 1997).

Trunk control is dependent on adequate sensory feedback and muscular control, as the passive stiffness of the spine's ligaments and intervertebral discs alone is insufficient to maintain trunk stability (Bergmark, 1989; Crisco and Panjabi, 1991). Previous studies have shown that upright stance control is attained by sensory integration mechanisms that generate corrective torques based on a combination of proprioceptive reflexes with short delays (20–50 ms) and corrective responses based on integrated information from proprioceptive, visual and vestibular systems with relatively long feedback delays (150–200 ms) (Cenciarini and Peterka, 2006; Maurer et al., 2006; Peterka, 2002). However, very little research has focused on the sensory information used in trunk control.

Vision contributes to trunk control with anticipatory feed-forward information (de Santiago et al., 2013; Krishnan and Aruin, 2011; Mohapatra and Aruin, 2013; Mohapatra et al., 2012) and/or reactive feedback information. Very little work has been done to investigate the influence of visual information on the reactive control of trunk posture. Goodworth and Peterka (2009) showed small but significant effects on trunk control of a tilting visual field during unpredictable pelvic-tilt perturbations, while the long-latency component (including visual information) of their model generated twice the amount of corrective torque compared to any other feedback component. In contrast, Moorhouse and Granata (2007) have suggested that trunk control under unpredictable external perturbations is mainly attributable to proprioceptive reflexes. Similarly, van Drunen et al. (2013) have shown that a model with intrinsic stiffness and damping and proprioceptive reflexes only (no visual and/or vestibular components) was able to describe the dynamic behavior of the trunk during dynamic perturbations. Therefore, trunk control may be different

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from stance control and a more detailed analysis of the contribution of sensory modalities with long (e.g. visual) and short (e.g. muscle-spindle) delays to trunk stabilization is needed.

Muscle spindles have an important influence on trunk control, through feedback of position and movement of the trunk. Muscle spindle contributions may depend on the posture of the lumbar spine. For example, sitting with a flexed spine lengthens the lumbar muscles, which affects the information from muscle spindles, and in turn, may affect reflex activity (Granata and Rogers, 2007; O'Sullivan et al., 2002). Furthermore, the passive stiffness of the trunk increases with flexion, lateral flexion and extension (McGill et al., 1994). Therefore, changing lumbar posture might influence the stability of the trunk and the contribution of intrinsic and reflexive mechanisms.

The goal of this study was to determine the effects of vision and lumbar posture on trunk neuromuscular control. Based on the modeling work of van Drunen et al. (2013), it was hypothesized that visual information has no effect on trunk control during small-amplitude trunk perturbations. Furthermore, it was hypothesized that posture would affect trunk control and change the relative contributions of intrinsic and reflexive mechanisms.

2. Methods

2.1. Subjects

The study was approved by the ethical committee of the Faculty of Human Movement Sciences of the VU University Amsterdam. Fourteen healthy subjects participated in the experiment (5 males, age range 22–28 years, mean mass: 74 kg ($\pm$13 kg)). All subjects gave their informed consent prior to the experiment. Subjects reported no low back pain in the year prior to the experiment and did not suffer from any visual impairments or neurological disorders that could affect balance control.

2.2. Experimental setup

Subjects were positioned in a kneeling-seated posture, while the pelvis was fixed to reduce pelvic motion (Fig. 1). During the trials, subjects were requested to keep their arms crossed in front of their chest. A force perturbation was applied at the level of the spinous process of T10 in ventral direction. For comfort and better force transmission, a thermoplastic patch (5×5 cm^2) was placed between the pushing rod and the back of the subject. During all trials, subjects were instructed to minimize flexion/extension and lateral flexion excursions and thus resist the force perturbation as good as possible. Each subject performed a total of 4 conditions: neutral posture-eyes open, neutral posture-eyes closed, flexion posture-eyes closed and extension posture-eyes closed. During the flexion and extension posture trials, the pelvis was rotated maximally in the posterior (flexion) or anterior (extension) direction while the trunk was kept upright, resulting in a lumbar flexion of 22.9° ($\pm$1.7°) and lumbar extension of 19.5° ($\pm$5.7°) compared to neutral posture. Each condition was repeated four times, giving a total of 16 trials per subject.

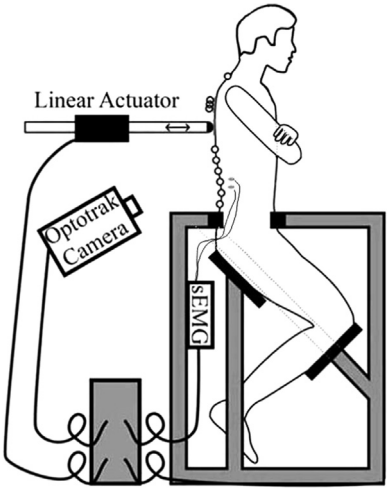

Fig. 1. The experimental setup.

2.3. Force perturbation

As the pushing rod was not attached to the subject, a 60 N preload was applied to maintain contact. Superimposed on the preload, a dynamic disturbance with a 35 N amplitude was applied (Fig. 2, second panel) as described by van Drunen et al. (2013). The dynamic disturbance ($F_p(t)$) was a crested multi-sine (Pintelon and Schoukens, 2001) of 20 s duration containing 18 logarithmically spaced frequency pairs with a bandwidth ranging from 0.2 to 15 Hz (Fig. 2, top panel). To reduce adaptive behavior to high frequency content, the power above 4 Hz was reduced to 40% (Mugge et al., 2007). Since the perturbation was perceived as random by the subjects, no feed-forward or voluntary activation was expected to occur in relation to the perturbation. Each force perturbation consisted of a 3 s ramp force increase to 60 N preload, a 2 s static preload, the last 5 s of the disturbance (as a start-up to reduce transient behavior) and twice the 20 s dynamic disturbance giving a total of 50 s per run.

2.4. Data recording and processing

The kinematics of the lumbar spine (L1–L5), the thorax (cluster of 3 markers at T6) and the pelvis (cluster of 3 markers at the sacrum) were measured using 3D motion tracking at 100 Hz (Optotrak3020, Northern Digital Inc, Canada). The actuator displacement ($x_A(t)$) and contact force ($F_c(t)$) between the rod and the subject were measured at 2000 Hz (Servotube position sensor & Force sensor FS6-500, AMTI, USA). Preliminary kinematic analysis revealed that rotation occurred both at the level of the lumbar spinal column and at the level of the pelvis. This indicated that the pelvic restraint was not able to completely eliminate movement of the pelvis. However, despite movement of the pelvis, all subjects showed substantial movement in the spine and could therefore be included for further analysis. Since the kinematic analysis indicated that an effective low-back bending rotation point, necessary to define rotations, was not well defined and inconsistent over subjects and tasks, trunk kinematics were described in terms of translation of the pushing rod.

Activity of the lumbar part of the M. Longissimus ($e(t)$) was recorded bilaterally at 2048 Hz with surface electromyography (sEMG; REFA, TMSi, the Netherlands). The electrodes were placed 3 cm lateral to the space between the spinous processes of L3 and L4. The M. Longissimus was chosen given a high coherence between its activity and trunk displacement (van Drunen et al., 2013). The EMG signals were digitally high-pass filtered at 250 Hz (first order, zero-phase) (Staudenmann et al., 2007), rectified and scaled to maximal voluntary contraction (MVC) level. The MVC was determined with 2 maximal extension contractions at the end of the experiment. For each contraction, the subject was instructed to build-up to a maximal contraction in 5 s and holds the contraction for 3 s against manual resistance provided by the experimenter. The EMG during the 3 s plateau was averaged and the highest value of both contractions was used as MVC. To test whether differences in reflex activity between the different conditions can be explained by an altered level of muscle pre-activation, the normalized EMG amplitude was calculated over the 2 s static preload (60 N) preceding the dynamic disturbance. Finally, since the M. Longissimus counteracted the perturbation, the muscle activity during the perturbation was expressed as negative.

2.5. System identification

Closed-loop identification (Schouten et al., 2008; van der Helm et al., 2002; van Drunen et al., 2013) was used to determine the trunk translational admittance ($H_{adm}(f)$) and reflexes)$H_{emg}(f)$) as frequency response functions (FRFs). The admittance describes the actuator displacement ($x_A(t)$) as a function of contact force ($F_c(t)$), whereas the reflexes describe the EMG ($e(t)$) as a function of actuator displacement ($x_A(t)$). Because the subjects interacted with the actuator, FRFs were estimated using closed loop methods.

$$H_{adm}(f) = \frac{S_{F_p x_A}(f)}{S_{F_p F_c}(f)}; \quad H_{emg}(f) = \frac{S_{F_p e}(f)}{S_{F_p x_A}(f)} \tag{1}$$

with $S_{F_p x_A}(f)$, $S_{F_p F_c}(f)$ and $S_{F_p e}(f)$ representing the estimated cross-spectral densities between the Fourier transformed force-perturbation ($F_p(f)$) and actuator displacement ($x_A(f)$),contact force ($F_c(f)$) and EMG ($e(f)$) respectively.

The cross-spectral densities were only calculated at the frequencies that contained power in the force perturbation. To reduce noise and improve the estimate, the cross-spectra were averaged across the 4 trials per condition, the two 20 s time segments (dynamic disturbance) and across the 2 adjacent frequency points (Jenkins and Watts, 1969). Finally, the cross-spectra between force perturbation and EMG were averaged across the left and right muscles. The coherence of the admittance (γ^2_{adm}) and reflexes (γ^2_{emg}) was calculated as follows:

$$\gamma^2_{adm}(f) = \frac{|S_{F_p x_A}(f)|^2}{S_{F_p F_p}(f) S_{x_A x_A}(f)}; \quad \gamma^2_{emg}(f) = \frac{|S_{F_p e}(f)|^2}{S_{F_p F_p}(f) S_{ee}(f)} \tag{2}$$

Coherence ranges from zero to one, where one reflects a perfect, noise-free relation between input and output. Since the spectral densities were averaged across 16 points, a coherence larger than 0.18 is considered significant at the $p < 0.05$ level (Halliday et al., 1995). Therefore, all frequency points with a subject-averaged coherence of 0.18 or larger were included for further analysis.

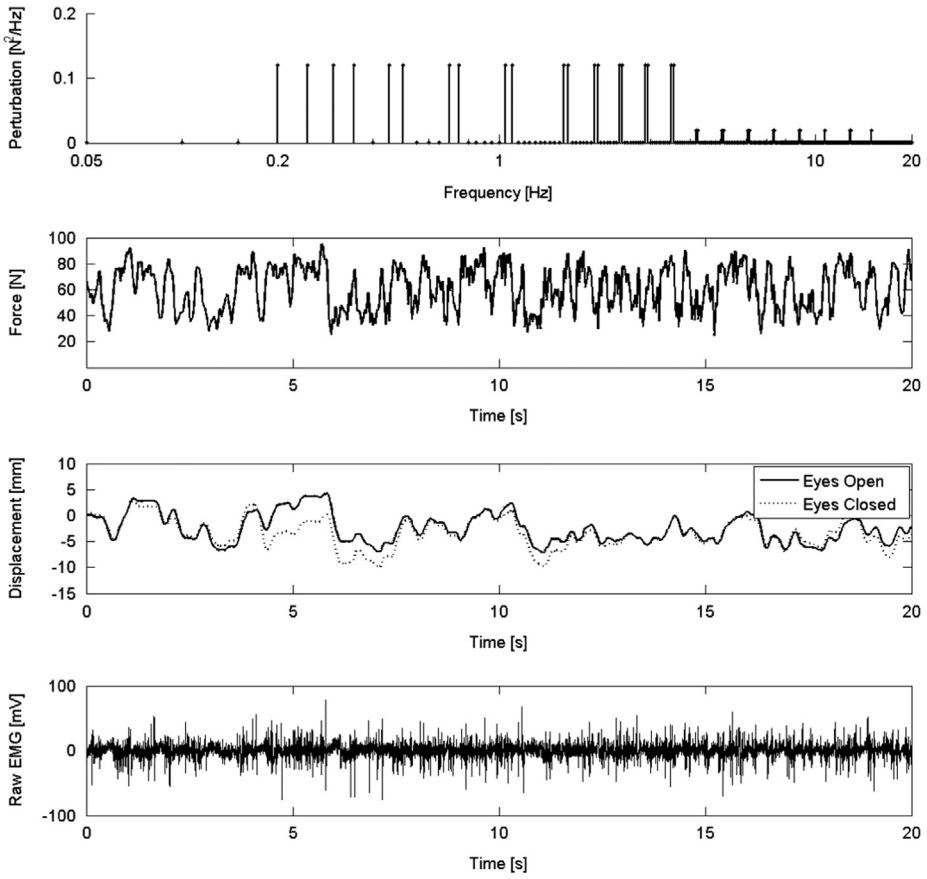

Fig. 2. The power spectrum (top panel) and time signal (second panel) of the applied force perturbation, a typical example of the measured displacement during an eyes-open and eyes-closed trial (third panel) and a typical example of the measured EMG during an eyes-open trial (bottom panel).

2.6. Statistics

The gains of the admittance and reflexes were log-transformed to satisfy the assumption of normality. Sphericity was checked using Mauchly's test. If the assumption of sphericity was violated, a Greenhouse–Geisser correction was used (Girden, 1992). Partial Eta Squared (η_p^2) was used as a measure of effect size. To investigate whether there was a significant difference in the admittance or reflexes between the eyes open and eyes closed condition, a 2 factor (condition [2] × frequency [18]) repeated measures ANOVA was performed on the gain of the admittance ($H_{adm}(f)$) and also on the gain of the reflexes)$H_{emg}(f)$). Furthermore, to investigate the differences between the neutral, flexion and extension conditions, another 2 factor (condition [3] × frequency [18]) repeated measures ANOVA was performed. Significant interaction effects were followed up by one-way repeated measures ANOVA's and significant main effects were followed up by Bonferroni corrected pair-wise comparisons.

Furthermore, to test if there was a significant difference between the levels of muscle pre-activation, a paired-samples *t*-test was performed on the EMG amplitude of the eyes-open and eyes-closed condition, and a one-factor repeated measures ANOVA was performed on the EMG amplitude of the neutral, flexed posture and extended posture conditions. A significant main effect was followed up by Bonferroni corrected pair-wise comparisons. Effects were considered significant when the corrected $p < 0.05$.

3. Results

A typical example of the measured position of one subject during the eyes-open and eyes-closed trials shows that the displacement corresponds with the force imposed (Fig. 2).

The trunk stabilizing behavior is described by the FRF's of the admittance and reflexes (Figs. 3 and 4), while high coherences indicate good input-output correlations. The subject-averaged coherence always exceeded the 0.05 probability level of 0.18 and therefore all data were used for further analysis. The FRF of the admittance resembles a combination of a second-order system (i.e., a mass-spring-damper system) and reflexive responses (c.f. Schouten and van der Helm, 2008). The behavior at high frequencies (> 2 Hz) is predominantly determined by the mass of the trunk and contact dynamics. The low-frequency behavior (< 1 Hz) is a reflection of the intrinsic stiffness and reflexes. The intermediate frequencies are also mainly determined by the reflexes and intrinsic damping. The FRF of the reflexes indicates the presence of position feedback (flat gain and -180° phase lag at lower frequencies), velocity feedback ($+1$ gain slope and -90° phase lag at intermediate frequencies) and acceleration and/or force feedback (second-order ramp-up at the high frequencies).

In line with the resemblance of the eyes open and eyes closed conditions in Fig. 2, no significant effects of vision were found in the gain of the admittance and reflexes (Table 1, Fig. 3). Furthermore, there was no significant difference between the baseline EMG of the eyes-open ($23 \pm 13\%$ of MVC) and eyes-closed ($26 \pm 20\%$ of MVC) condition ($p = 0.309$). Therefore, the hypothesis was confirmed that visual information does not contribute to stabilization of the trunk in the present task.

For the influence of posture, two significant effects were found (Table 2). The flexed posture condition led to a significantly lower admittance gain compared to the neutral posture ($p = 0.028$, 95% CI [−0.237, −0.013]) (Fig. 4, left top panel). The significant interaction indicated that this difference occurred mainly at the low (0.3–1 Hz) and high (7–15 Hz) frequencies. There were no significant differences in the admittance for the extended posture condition. For the reflexes, a lower gain in the flexed posture, compared to neutral, was visible but this failed to reach statistical significance ($p = 0.053$, 95% CI [−1.177, −0.008]) (Fig. 4, right top panel). The flexed posture condition did lead to a significantly lower reflex gain compared to

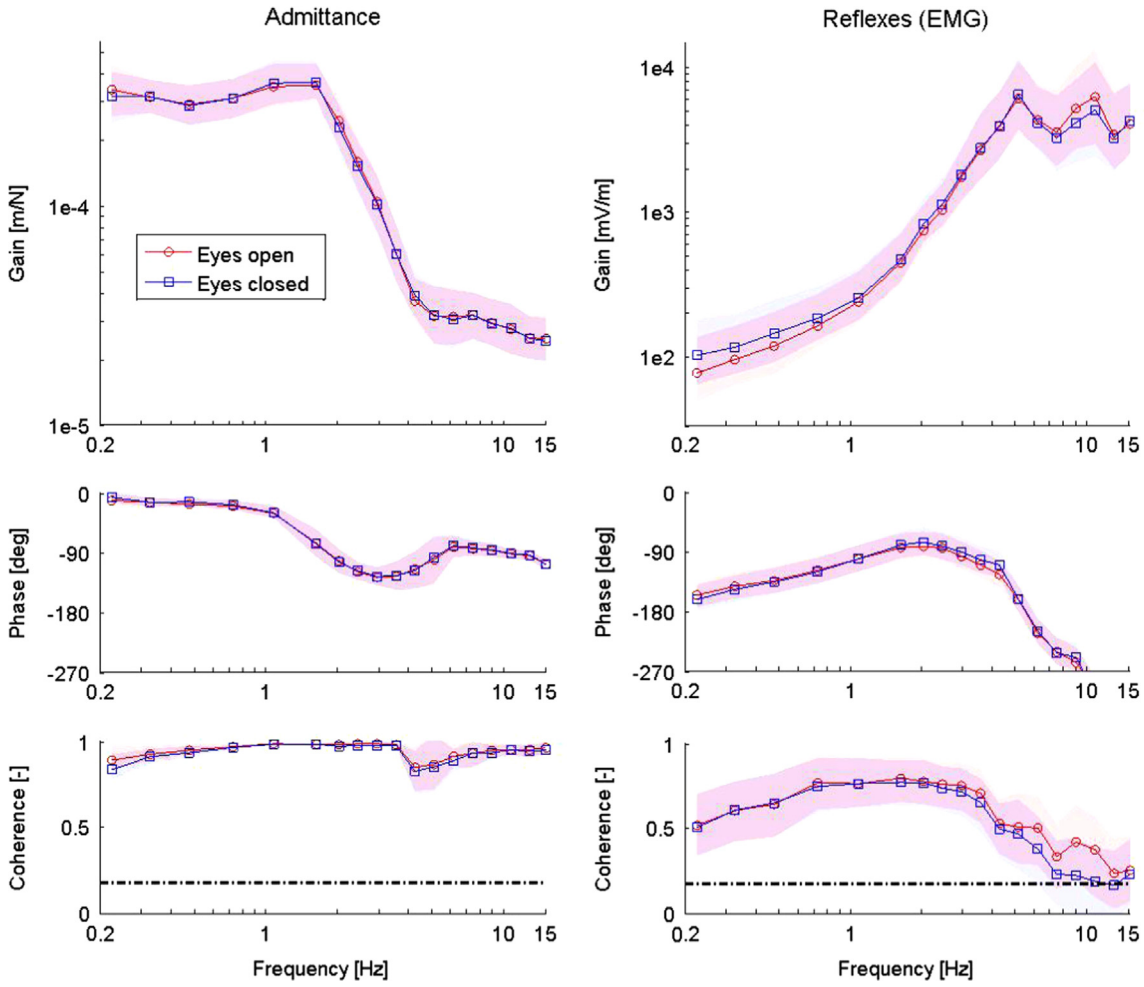

Fig. 3. Frequency response function for the eyes-open (O) and eyes-closed (□) condition averaged across all subjects. The shaded area represents one standard deviation. The dashed line in the lower plots represents the significance level for coherence.

the extended posture condition ($p=0.013$, 95% CI [−1.678, −0.193]). The changes in the reflex gains coincided with changes in muscle pre-activation as the EMG amplitude during the flexion posture ($15 \pm 12\%$ of MVC) was significantly lower ($p=0.005$, 95% CI [−18.2, −3.4]) compared to the neutral condition ($26 \pm 20\%$ of MVC). Therefore, the second hypothesis was confirmed that posture would influence the intrinsic and reflexive contributions to trunk control.

4. Discussion

The purpose of the present study was to determine the effect of vision and sitting posture on trunk neuromuscular control. The results showed that having the eyes open had no effect on trunk neuromuscular control, thereby confirming that visual information does not contribute to stabilization of the trunk during small-amplitude trunk perturbations. Changing the posture of the lumbar spine did have an effect on the intrinsic and reflexive contributions to trunk control. In particular, flexing the lumbar spine led to a lower admittance and a lower contribution of reflexes.

The absence of an effect of vision on trunk control in the present experiment, is in contrast with the results of Goodworth and Peterka (2009). They found that manipulating visual information by tilting the visual field, had an effect on the motor control of the spine. There might be several explanations for this disparity: direction of perturbation, perturbation type/experimental setup, and visual flow amplitudes.

In the present experiment, the perturbation was in the anterior–posterior direction, while Goodworth and Peterka (2009) perturbed in the medio-lateral direction. The visual flow in the medio-lateral direction may provide more information on trunk orientation and consequently have a stronger effect on trunk control. However, when subjects are perturbed in anterior–posterior direction by sitting on a moving platform, an effect of vision is observed (van Drunen et al., submitted for publication). Therefore, perturbation direction cannot be the only explanation.

More likely, the difference in the results might be explained by the experimental setup and perturbation type. In the present study, the trunk (and therefore head) position in space could be controlled by the visual feedback, as well as by the proprioceptive feedback, as both feedback mechanisms would counteract a displacement of the trunk/head in space. During the perturbations applied by Goodworth and Peterka (2009) and van Drunen et al. (submitted for publication), only visual feedback is appropriate to maintain the trunk/head position in space, while proprioceptive trunk feedback minimizes lumbar bending and thus aggravates the trunk/head displacements in space. Therefore, in the present experiment, a trade-off between visual and proprioceptive information can exist whereas in the experiments of Goodworth and Peterka (2009) and van Drunen et al. (submitted for publication), both sources provide unique information.

Finally, the results of Goodworth and Peterka (2009) showed that the effect of visual information is dependent on the amplitude of visual flow. The contribution of vision increased with larger

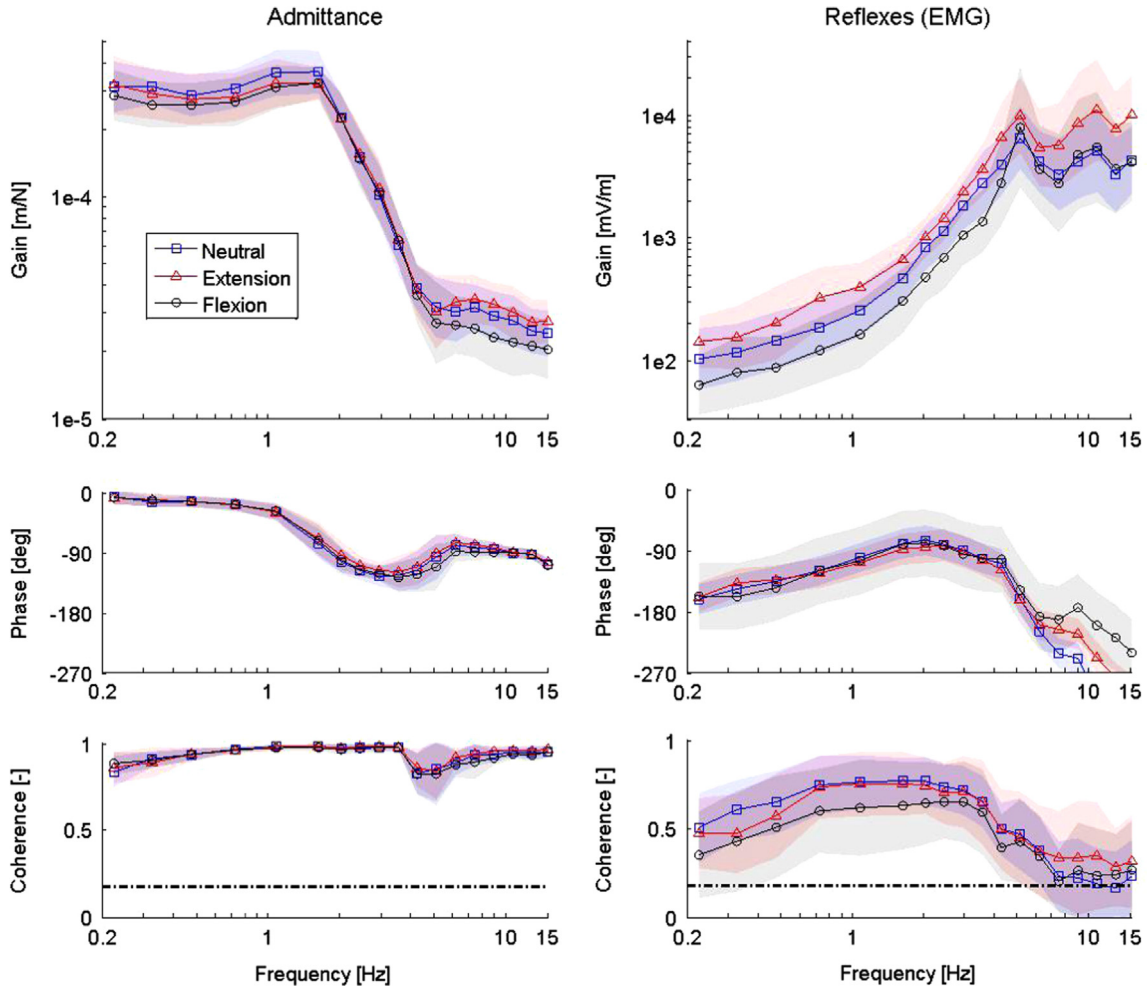

Fig. 4. Frequency response functions of the neutral (□), extended (Δ) and flexed (O) posture conditions averaged across all subjects. The shaded area represents one standard deviation. The dashed line in the lower plots represents the significance level for coherence.

Table 1
Main and interaction effects of the ANOVA's for vision on the gain of the admittance and reflexes. η_p^2 was used as a measure of effect size.

Admittance					
	F	***df***	***P***	η_p^2	**Pairwise comparisons**
Condition	0.126	1, 12	0.729	–	–
Condition × frequency*	1.044	3.7, 44.2	0.392	–	–
Reflexes					
Condition	0.326	1, 12	0.579	–	–
Condition × frequency*	2.076	2.7, 33	0.127	–	–

* Greenhouse–Geisser correction due to a violation of the assumption of sphericity.

amplitudes of visual manipulations. Therefore, in the present experiment, the amplitude of the perturbation might have led to only small displacements of the upper body/head in space and the resulting visual flow might have been too small to excite an effect of visual information on trunk control. However, this still answers a relevant question for many activities of daily life (e.g., standing, sitting, desk work, etc.) in which only small upper body/head motion occurs.

Compared to the neutral posture, flexing the lumbar spine led to a decrease in the gain of the admittance, indicating more resistance to the perturbation. Since a decrease in reflex gain and pre-activation of the M. Longissimus was observed, the higher resistance could not be the result of higher muscle activation or a higher co-contraction level. However, there are indications that flexing the spine puts the muscles in a more optimal range of the force-length relationship (Raschke and Chaffin, 1996). Therefore, the same torque could have been generated with decreased activation. Furthermore, the increased flexion may have led to an increase in passive tissue stiffness (McGill et al., 1994) which also could have compensated for the decreased muscle activation (both reflex activity and co-contraction). The flexion–relaxation phenomenon might explain the decreased muscle baseline and reflex activity (Rogers and Granata, 2006; Solomonow et al., 1999). Finally, the increased passive tissue stiffness itself, may have led to a lower reflex gain.

Several limitations need to be discussed. First, only a limited number of subjects participated in this experiment, which could have limited power to detect differences between conditions. However, the results do not indicate any non-significant trends. Second, there was a lack of complete pelvis fixation which, in combination with movement at the SI-joints, allowed the pelvis to contribute to the motion of the trunk. However, this contribution was consistent for all subjects and all trials and therefore did not influence any differences between conditions.

In conclusion, visual information does not seem to play a role in controlling trunk posture under small-amplitude anterior–posterior torso perturbations. In contrast, posture does affect trunk control, through changes in the intrinsic stiffness and proprioceptive reflex activation.

Table 2
Main and interaction effects of the ANOVA's for posture on the gain of the admittance and reflexes. η_p^2 was used as a measure of effect size.

Admittance					
	F	***df***	***P***	η_p^2	**Pairwise comparisons**
Condition*	7.119	1.4, 16.4	0.011	0.372	Flexion < neutral ($p=0.028$)
Condition × frequency*	2.853	4.1, 49.3	0.032	0.192	Flexion < neutral @ 0.3, 0.7, 1, 7, 9, 11, 13, 15 Hz
Reflexes					
Condition	9.011	2, 24	0.001	0.429	Flexion < Extension ($p=0.013$); Flexion < Neutral ($p=0.053$)
Condition × frequency*	0.884	6.3, 75.9	0.515	–	–

* Greenhouse–Geisser correction due to a violation of the assumption of sphericity.

Conflict of interest statement

The authors declare that no conflicts of interest were associated with this study.

Acknowledgments

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